DEVELOPMENT OF NEW WHIPLASH PREVENTION SEAT

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ABSTRACT

Whiplash, or soft tissue cervical injury, is a common injury incurred by occupants of passenger cars in rear-end collisions. Despite much investigation into the cause of such injury, no single mechanism describes it completely. Proposed criteria focus on the relative motions of the head and the thorax, while few case studies have been made on the motions of the cervical vertebrae. Recently, the human body finite element model called "THUMS"(Total HUman Models for Safety) and the use of X-ray cineradiography devices by volunteers have accelerated the investigation into the motions of the cervical vertebrae.

Seats have been developed that are specially designed to reduce impact on the neck in rear-end collisions by simultaneously restraining the head and body of the occupant and controlling their motion relative to each other. We have developed a seat that also reduces local strain of the neck by preventing the rotation of the head, and that uniformly distributing the loads on the cervical vertebrae.

A finite element model was used to simulate rear-end collisions under the same conditions as sled tests using a BIO-RID II dummy, with a THUMS human model placed on our newly developed seat. Prior to the simulation, the validity of the THUMS was investigated by comparing its head and neck motions with those in experiments. The validated THUMS predicted a reduction of local strain in the neck on the newly developed seat.

Having succeeded in reducing both the injury values to the dummy and the local strain of the neck of the THUMS, we predict that our new seat design to help reduce whiplash injury.

1. INTRODUCTION

Soft tissue cervical injury (whiplash) is a common injury resulting from rear-end collisions in passenger cars. As shown in Figure 1 [1-2], rear-end collisions account for about 50% of accidents resulting in injury, although only a small number of them are fatal. About 80% of injured occupants suffer neck injury, and the reduction of whiplash is,

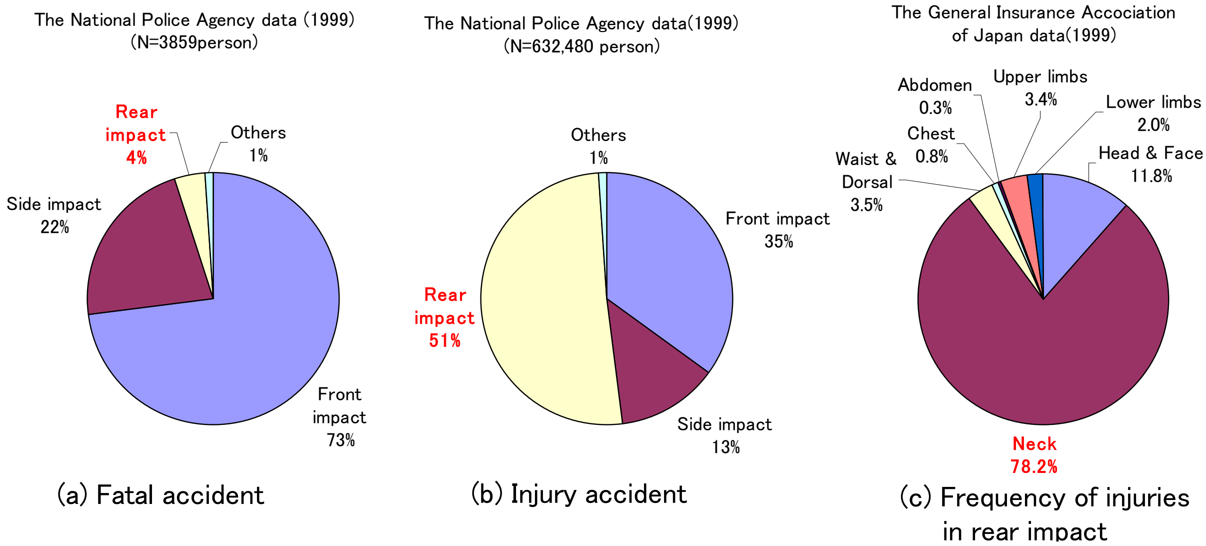


Fig.1 Realities of rear-end collision.

therefore, an important issue.

Krafft, et al. [4] analyzed the acceleration pulses of actual collisions using an in-vehicle data recorder. Based on this data, acceleration pulses obtained from sled tests conducted by rating organizations, including Folksam, IIWPG and ADAC, were considered. So far, a triangular pulse sled of ΔV = 16km/h is the most widely adopted test. In this test, the equivalent collision velocity is 32 km/h for a vehicle-to-vehicle collision between two cars of the same weight, which represents only a 60th percentile collision in Japan, as shown in Figure 2 [3]. If collision velocities of up to 50 km/h are examined, as much as 90% of all such collisions can be represented. In other words, when converted into ΔV for a collision between two cars of the same weight, the ΔV = 25 km/h.

Various attempts have been made to clarify the injury mechanism of whiplash. However, whiplash injury cannot be described with only a single mechanism, and so a variety of criteria have been proposed. Bostron, et al. [5] have proposed a criterion called a neck injury criterion (NIC) based on the variation in the pressure of the spinal fluid within the cervical spinal canal. Schmitt, et al. [6] have proposed Nkm focusing on the shear force and bending moment of the upper neck, whereas

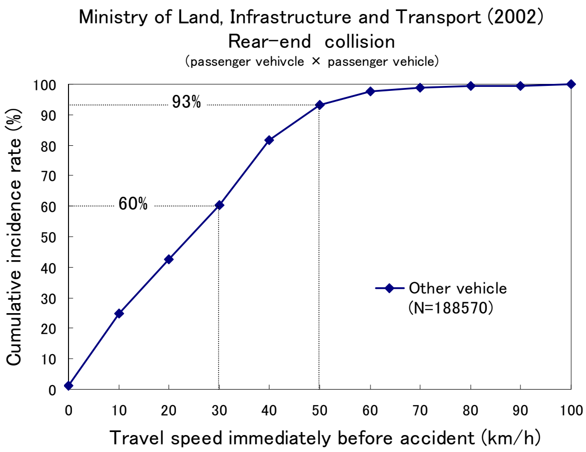


Fig.2 Cumulative incidence rate of travel speed immediately before accident.

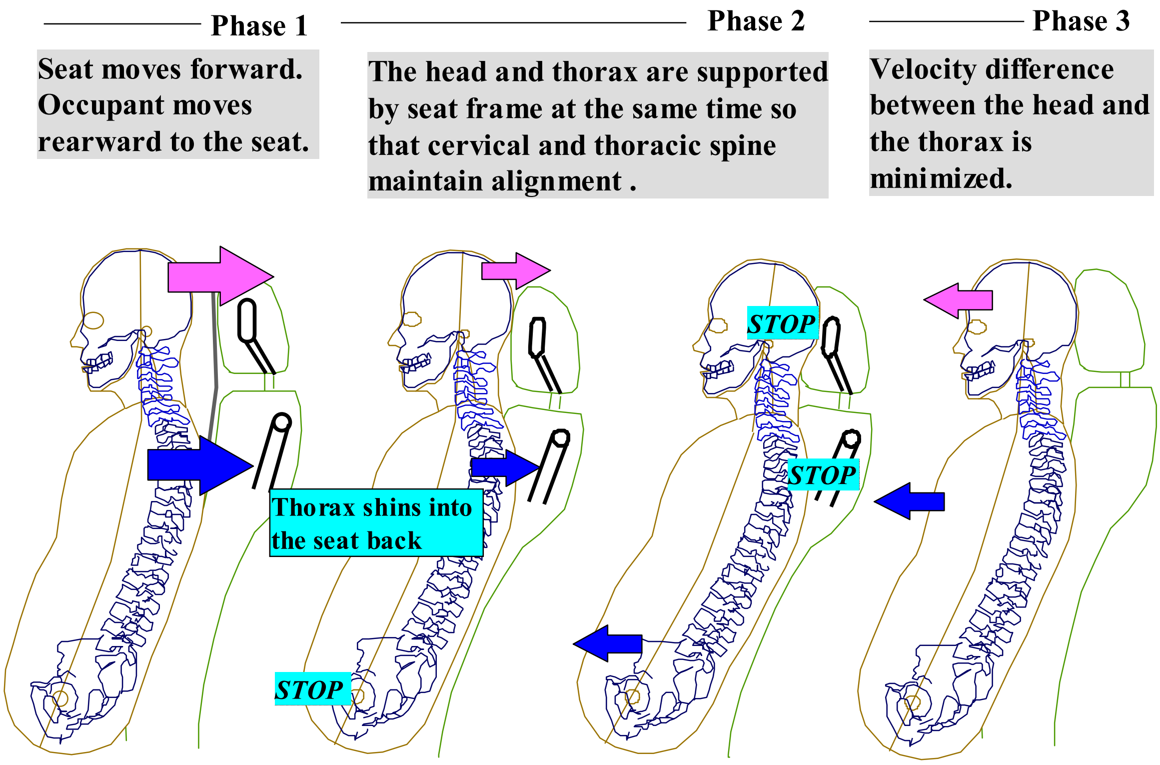


Fig.3 Whiplash injury lessening concept.

Heitplatz, et al. [7] have suggested a lower neck load index (LNL) that correlates with insurance claims for cervical vertebrae injuries, and emphasizes the shear force, axial force, and bending moment of the lower neck. Viano, et al. [8] have developed an advanced a neck displacement criterion (NDC) as a criterion for the movable range of the head and neck based on tests conducted on volunteers. Additionally, Panjabi, et al. [9] have presented IV-NIC for evaluating neck injury based on the ratio of it to the physiological limit rotating angle of the cervical vertebra joints.

Deng et al. [10] and Ono et al. [11] have analyzed the motions of the cervical vertebrae of corpses and of volunteers using X-ray cineradiography devices. On the other hand, Ejima et al. [12] and Hasegawa [13] have used human FE models to perform in-depth analyses of the stress and strain of the cervical vertebrae. Additionally, Lee et al. [14] have attributed cervical facet capsule distraction as a cause of neck pain by whiplash per their experiments on rats. Research into the mechanism of whiplash is thus shifting focus away from the relative motions of the head and the thorax and neck loads, toward the relationship of the local motions of the neck to whiplash.

We have developed a WIL seat [15] designed to reduce load on the neck based on a unique concept of the prevention of whiplash achieved by restraining the head and body of the occupant simultaneously and thereby controlling relative motions in a rear-end collision, as shown in Figure 3. Other companies have developed the shock absorption seat [16] and the active head restraint seat [17], both of which are intended to reduce loads on the neck. We have developed a seat that not only achieves the abovementioned objectives, but can also control local strain of the neck by preventing the rotation of the head and by uniformly distributing load on the cervical vertebrae. We also verified the effects of the seat through experiments using a BIO-RID II dummy and finite element (FE) analyses on a human FE model (THUMS).

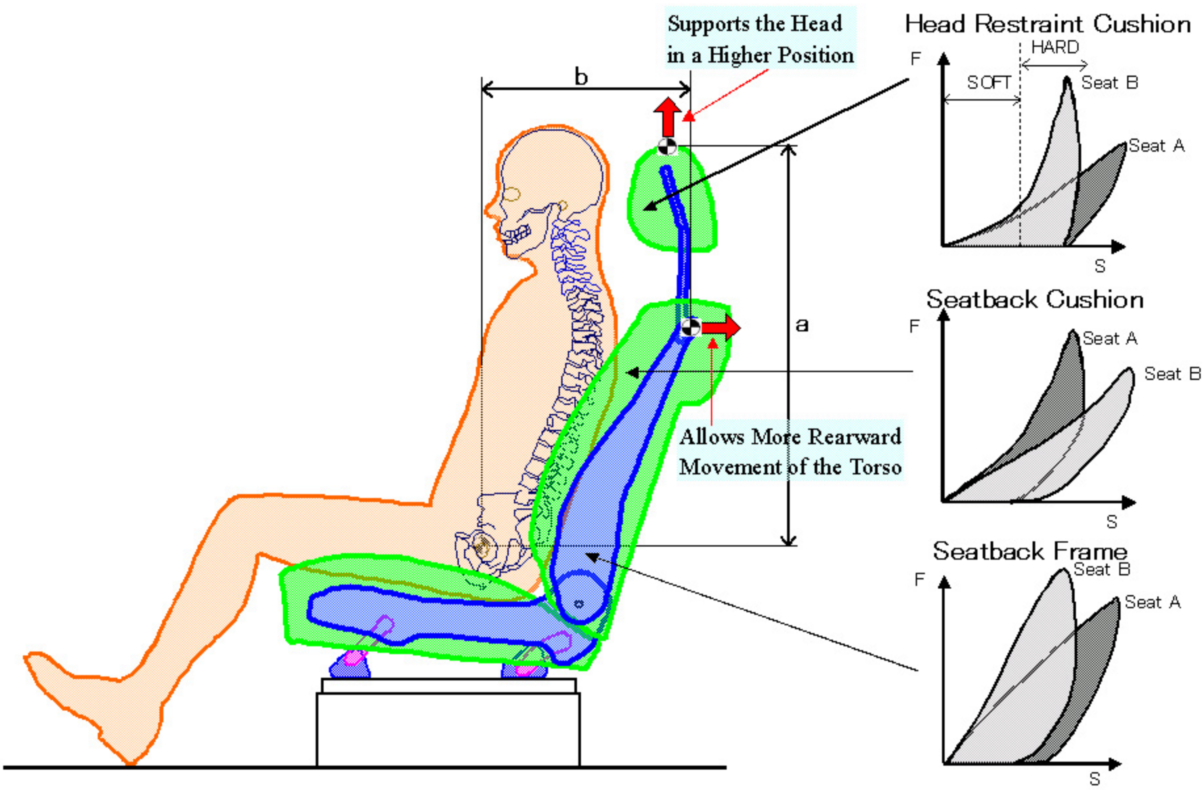


Fig.4 Comparison between a conventional seat (Seat A) and the newly developed seat (Seat B).

2. TEST METHODS

2.1 Test samples

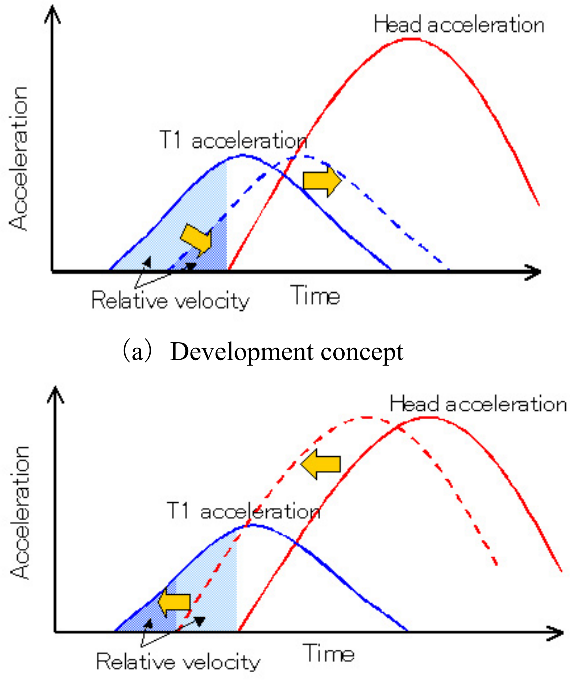
Figure 4 shows a comparison between a conventional seat (Seat A) and the newly developed seat (Seat B). One of the features of the new design is that the position (b) of the upper part of the frame and the stiffness of the seatback cushion were changed so that the thorax of the occupant sinks deeper into the seat in Phase 1 (until the head contacts the head restraint). This delays the onset of the rebounding motion of the thorax G (T1G) and reduces the velocity relative to head, as shown in Figure 5 (a). Similarly, the design of this newly developed seat (Seat B) aims to reduce the velocity relative to thorax G (T1G), but takes a different approach than the active head restraint, which causes the head G to rebound from the head restraint earlier as shown in Figure 5 (b).

We also reviewed the vertical position (a) of the head restraint and the F-S characteristic of the head restraint cushion to ensure restraint in Phase 2 (after the head contacts the head restraint). The balance of seat frame strength was also reconsidered, to provide reliable restraint performance at higher velocities up to ΔV = 25 km/h. These factors were changed with the aim of preventing the head of the occupant from extending over the head restraint even in a high-velocity rear-end collision by securely restraining the head at a higher position.

As mentioned earlier, currently there is no commonly accepted theory for the mechanism of whiplash at present, and a variety of criteria are proposed. In the following section, we verify the performance of the seat developed based on the abovementioned design through FE analyses using these proposed criteria and THUMS.

2.2 Test devices and conditions

An electrically controlled servo-hydraulic sled tester (Figure 6) was used to reliably generate free collision acceleration pulses. We conducted tests using a BIO-RID II (Figure 7) dummy, the most popular dummy for whiplash evaluation. The test method complies with the IIWPG test protocol [18]. Sled acceleration Pulses 1, 2, and 3 (shown in Figure 8) were used for testing. Pulse 1 is the most widely adopted acceleration pulse and a triangular pulse of ΔV = 16 km/h, tested by ADAC, Folksam and each



（a） Development concept

（b） Active head restraint system concept

Fig.5 Whiplash lesseing concept comparison.

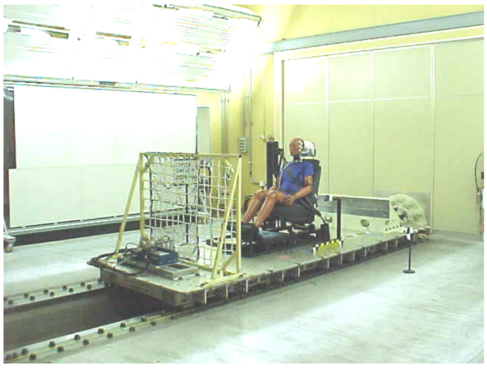


Fig.6 Electrically controlled servo-hydraulic sled tester.

Fig.7 BIO-RID II dummy.

IIWPG organization (IIHS and Thatcham). Since there is no standardized sled acceleration pulse of $\Delta V = 25$ km/h, we tested a triangular pulse of $\Delta V = 25$ km/h used by ADAC [19] for pulse 2, and a trapezoidal pulse of $\Delta V = 24$ km/h adopted by Folksam [20] for pulse 3.

3. RESULTS OF EXPERIMENTS AND DISCUSSION

This section discusses the results of conventional seat (Seat A) and the newly developed seat (Seat B), with sled acceleration pulses of pulses 1 to 3.

3.1 Discussion based on new criteria $T1G_{HRC}$, $T1V_{HRC}$ and $T1S_{HRC}$

Figure 9 shows the analysis results of thorax G (T1G). As shown in Figure 9 (a), there is no significant difference in $T1G_{max}$ between the conventional and new seats. However, the new seat produced lower values for T1G ($T1G_{HRC}$) until the head contacts the head restraint (T_{HRC}), for velocity variation $T1V_{HRC}$ (Expression 1) and for displacement $T1S_{HRC}$ (Expression 2) as shown in Figures 9 (b) to (d). This result indicates that the relative G, V, and S of the head and thorax are low in Phase 1 and head and thorax are more uniformly restrained. Figure 10 shows a comparison of head G

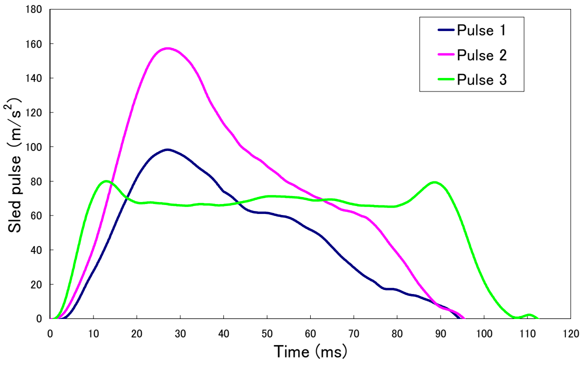
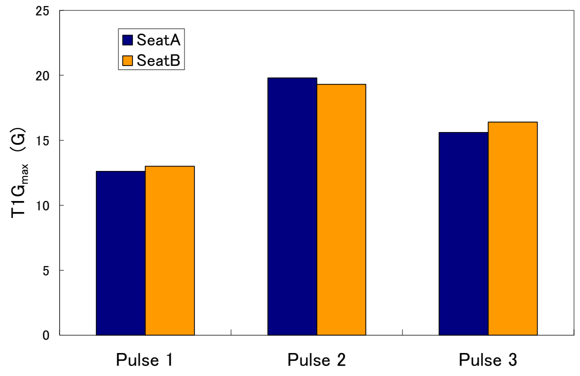
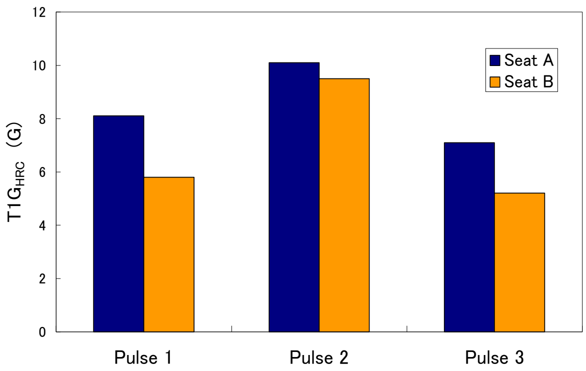


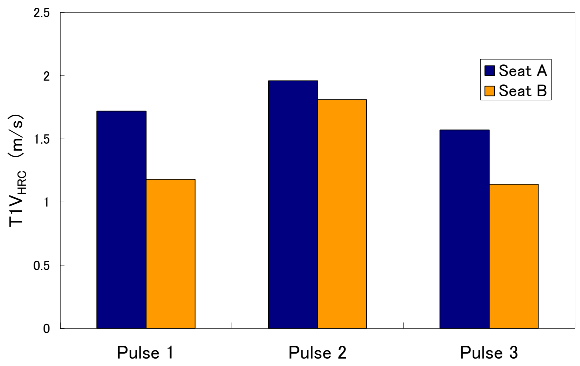
Fig.8 Test sled pulses.



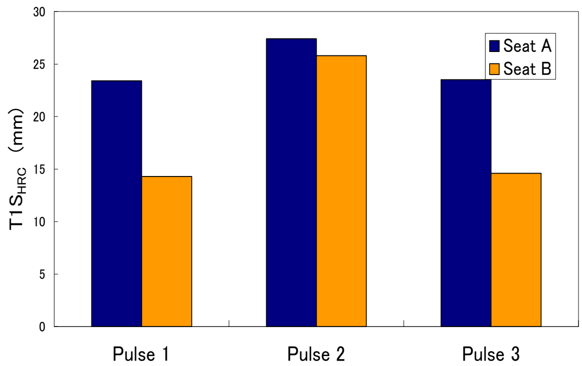
(a) $T1G_{max}$



(b) $T1G_{HRC}$

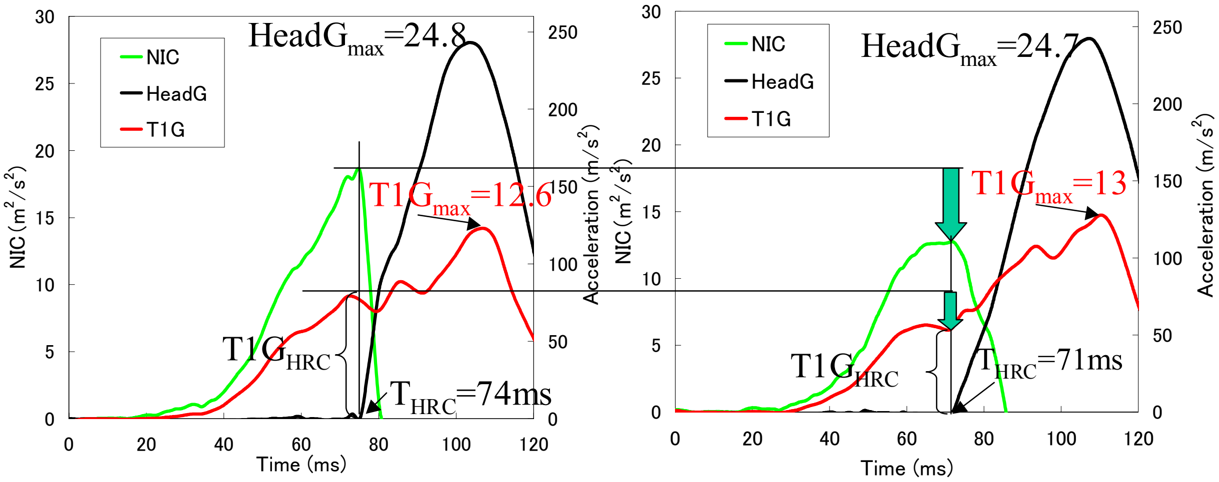


(c) $T1V_{HRC}$



(d) $T1S_{HRC}$

Fig.9 Examination with thorax G (T1G).



(a) Seat A (b) Seat B

Fig.10 Comparison of head G, thorax G, and NIC in Pulse 1.

and thorax G (T1G) in Phase 1 between the two seats. At maximum T1G, the head and the thorax are restrained, T1G(t) < HeadG(t), and the head is securely restrained and its extension restricted. Both seats had almost equal T_{HRC}, but $T1G_{HRC}$ (and $T1V_{HRC}$ and $T1S_{HRC}$) was lower in the new seat (Seat B). This indicates that the thorax sinks deeper into the seat in Phase 1 as intended.

$$T1V_{HRC} = \int_0^{T_{HRC}} T1G(t)dt \qquad (1)$$

$$T1S_{HRC} = \int_0^{T_{HRC}} \int_0^{T_{HRC}} T1G(t)dt^2 \qquad (2)$$

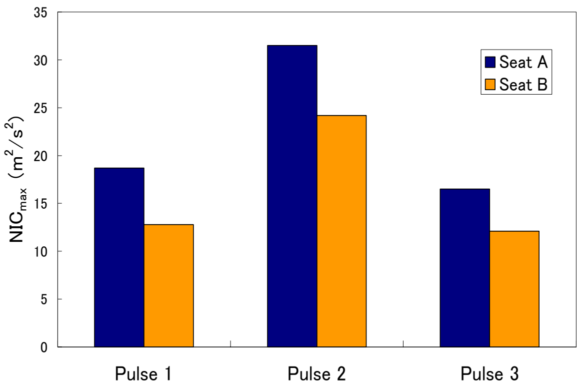


Fig.11 NIC examination.

3.2 Discussion based on conventional criteria

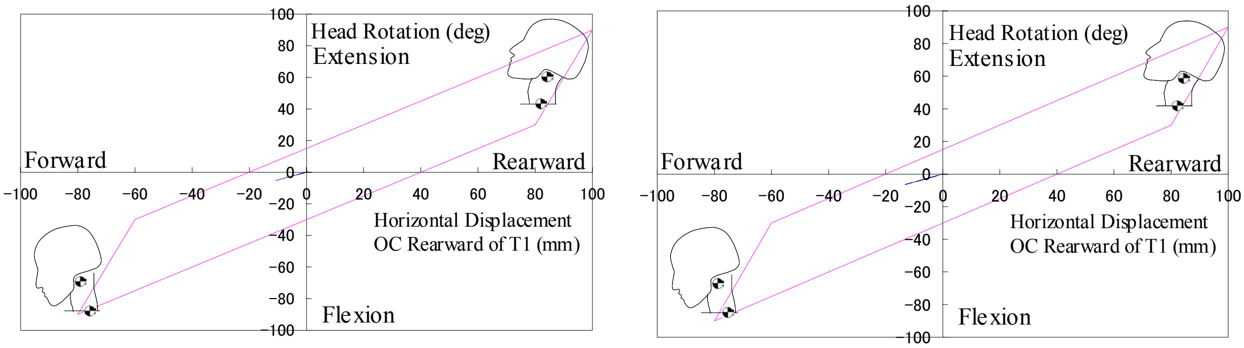
Figure 11 shows test results compiled based on the whiplash evaluation criterion NIC proposed by Bostron, et al. [5]. The NIC is a criterion for evaluating Phase 1 with emphasis placed on the relative motions of head G and thorax G (T1G) (Expression 3) and considered to be consistent with the new seat concept. NIC_{max} of the new seat (Seat B) was smaller than that of the conventional seat (Seat A) at the three sled pulses.

$$NIC(t) = 0.2 \cdot a_{rel}(t) + (V_{rel}(t))^2 \qquad (3)$$

where $a_{rel} = T1G(t) - HeadG(t)$

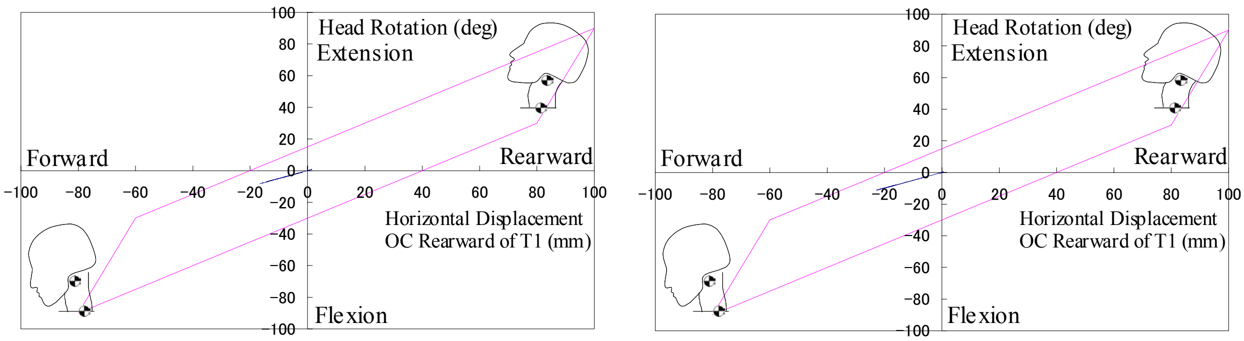
$$v_{rel} = \int a_{rel}(t)dt$$

Figure 12 shows the relationship between (a) horizontal displacement and head rotation at the NDC proposed by Viano, et al. [8] and between (b) horizontal displacement and vertical displacement. This criterion is considered to match the design concept of simultaneously restraining the head and the thorax and thereby preventing head rotation in Phase 2. Both seats securely prevent the vertical and rearward motions, rearward rotation of the head and the thorax, and are expected to provide a higher level of whiplash-reducing performance.



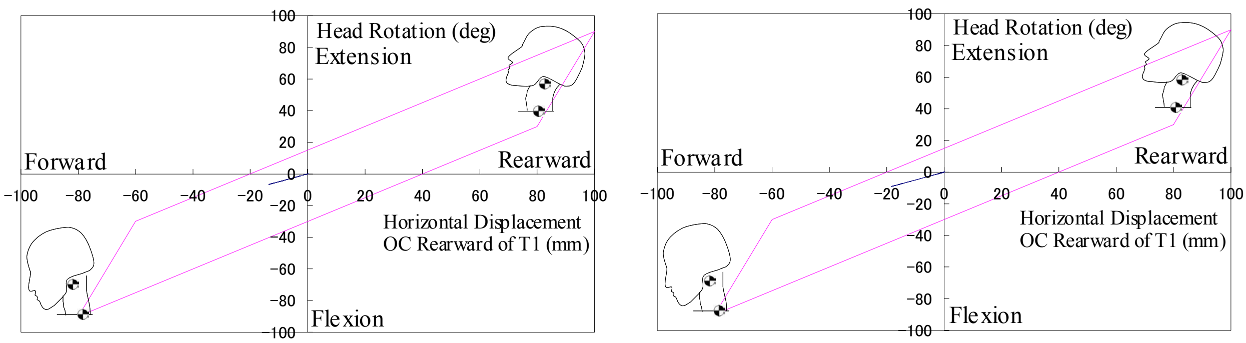
Seat A Seat B

(a) Pulse1 : ΔV=16km/h Triangular pulse



Seat A Seat B

(b) Pulse2 : ΔV=25km/h Triangular pulse

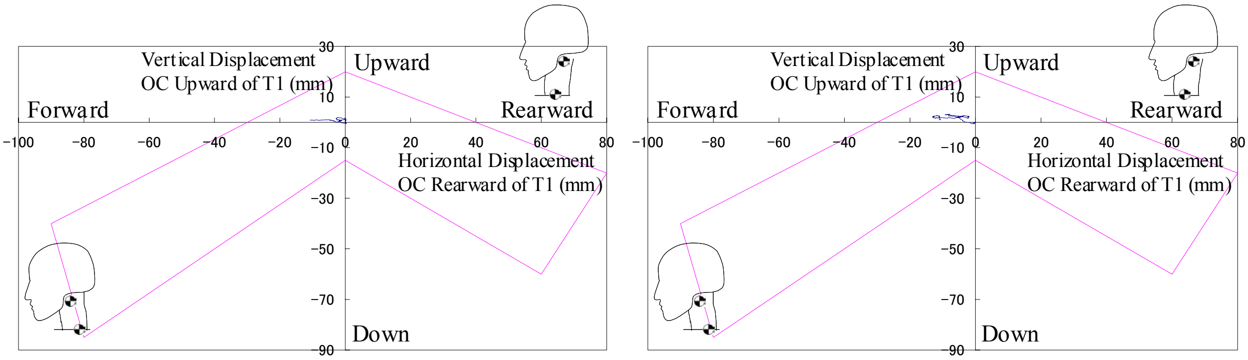


Seat A Seat B

(c) Pulse3 : ΔV=24km/h Trapezoidal pulse

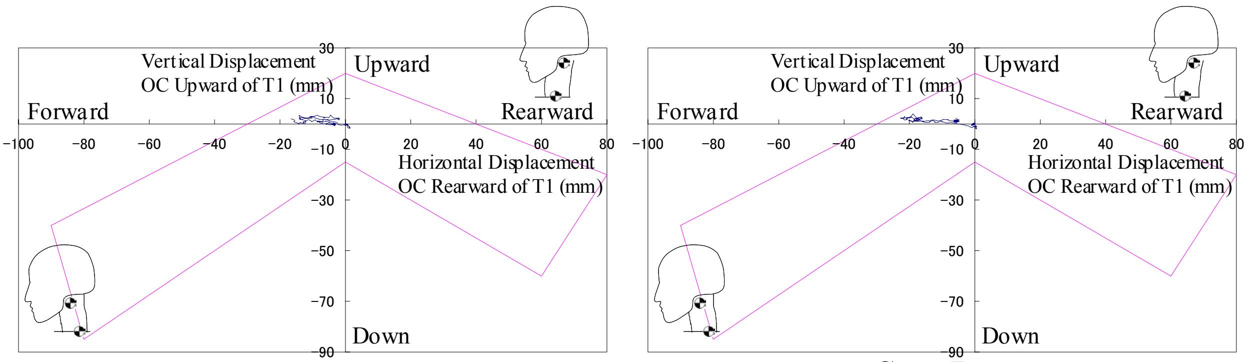
(a) Relationship between Horizontal Displacement and Head Rotation

Fig.12 NDC examination.



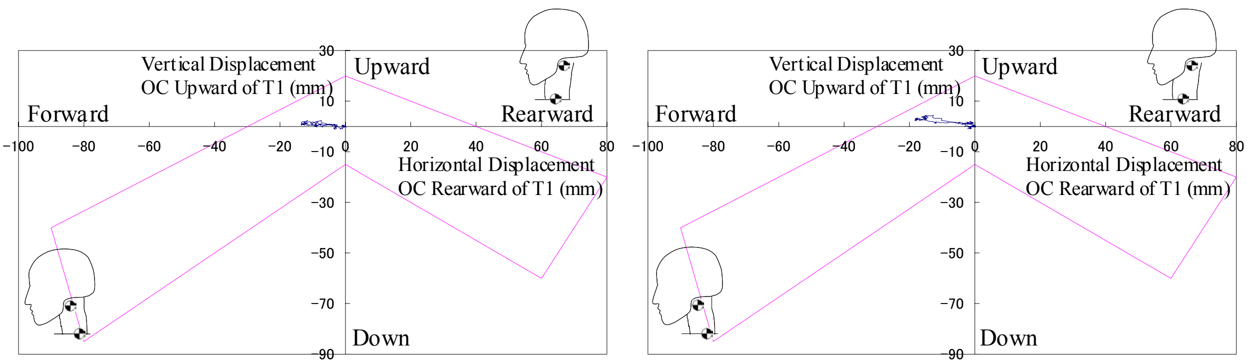
Seat A Seat B

(a) Pulse1 : Δ V=16km/h Triangular pulse



Seat A Seat B

(b) Pulse2 : Δ V=25km/h Triangular pulse



Seat A Seat B

(c) Pulse3 : Δ V=24km/h Trapezoidal pulse

(b) Relationship between Horizontal Displacement and Vertical Displacement

Fig.12 NDC examination.

Figure 13 shows the value of each component of the LNL (Expression 4) advanced by Heitplatz, et al. [7]. The LNL is a criterion focusing on the shear force, axial force, and bending moment of the lower neck in Phase 2. It draws attention as a criterion relating to the large angular variation and strain of the lower cervical vertebrae, such as C4-C5, C5-C6, and C6-C7, obtained from the results of the volunteer tests conducted by Sekizuka [15] and the results of the human FE analyses performed by Hasegawa [13]. The LNL indicates that the new seat (Seat B) produces lower values than the conventional seat (Seat A) at all three sled pulses and is, therefore, expected to yield a higher level of whiplash prevention.

Figure 14 shows the maximum value of lower neck extension moment My. This criterion was developed by focusing on the direct representation of load on the neck when it extends in Phase 2. This is also used in Expression 4 of the abovementioned LNL. The My_{max} of the new seat (Seat B) is lower than that of the conventional seat (Seat A) at all three sled pulses, suggesting that the former better prevents neck extension.

It can be seen in Figures 9 through 14 that the pulse shape exerts a significant effect on the injury value, for example there was a large difference in injury value between pulses 2 and 3 even though their ΔV is almost equal. On the whole, it was revealed that the injury value at pulse 2 was higher than that the value at pulse 3.

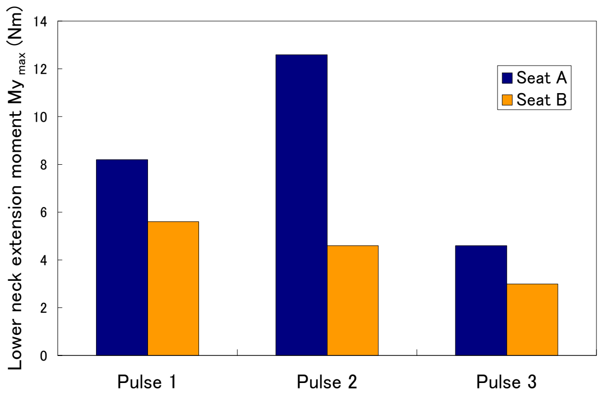
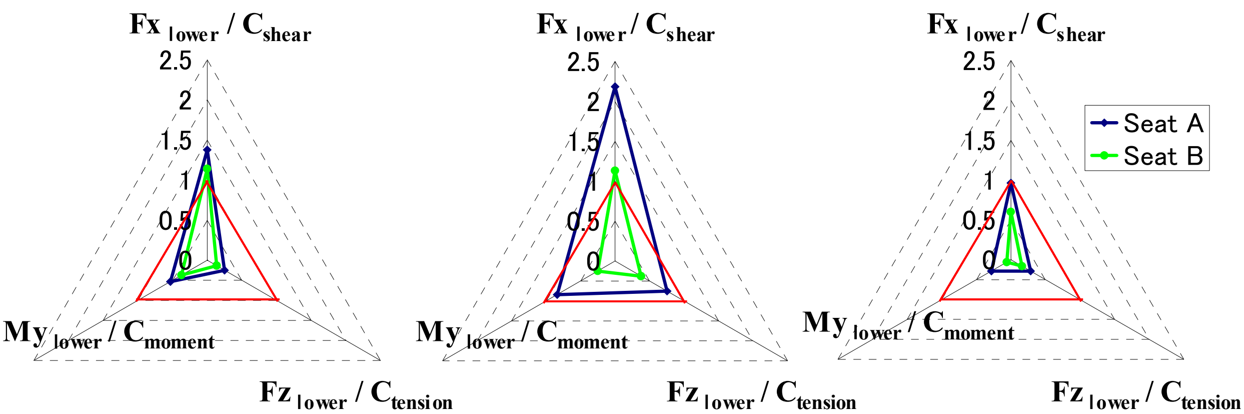


Fig.14 Lower neck extension moment My_{max} examination

$$\mathrm{LNL-index}(t)=\left|\frac{\sqrt{My_{lower}(t)^2+Mx_{lower}(t)^2}}{C_{moment}}\right|+\left|\frac{\sqrt{Fx_{lower}(t)^2+Fy_{lower}(t)^2}}{C_{shear}}\right|+\left|\frac{Fz_{lower}(t)}{C_{tension}}\right| \qquad (4)$$

Intercept values : $C_{moment}=15$, $C_{shear}=250$, $C_{tension}=900$

In BIO-RID II dummy : $Mx_{lower}(t)=0$, $Fy_{lower}(t)=0$



(a) Pulse1
ΔV=16km/h Triangular pulse

(b) Pulse2
ΔV=25km/h Triangular pulse

(c) Pulse3
ΔV=24km/h Trapezoidal pulse

Fig.13 LNL examination.

4. FE ANALYSES

4.1 FE model and its verification

To confirm the ability of the new seat to reduce loads on the human neck and to analyze how load on the neck occurs, we used a human FE model called a total human model for safety (THUMS) to simulate rear-end collisions. Figure 15 shows an enlargement of the entire body and neck of the THUMS-AM50 OCCUPANT Ver.1.63-050304. This model is an upgraded version of conventional THUMS passenger model v1.5, and was subjected to significant improvements such as the added ability to represent the structure of the spinal cord and the cervical vertebrae joints in detail, in order to accurately reproduce the motions of the neck and to evaluate the strain of the soft neck tissues in rear-end collisions. We also prepared a seat model representing a conventional seat structure and one combining newly developed structural design attributes (Figure 16), both of which were given the strength characteristics obtained from component tests (Figures 17 and 18). A THUMS v1.63 was placed in almost the same posture as the BIO-RID II dummy in both seat models, and rear-end collisions were simulated by inputting acceleration pulses equivalent to those used for the sled experiments.

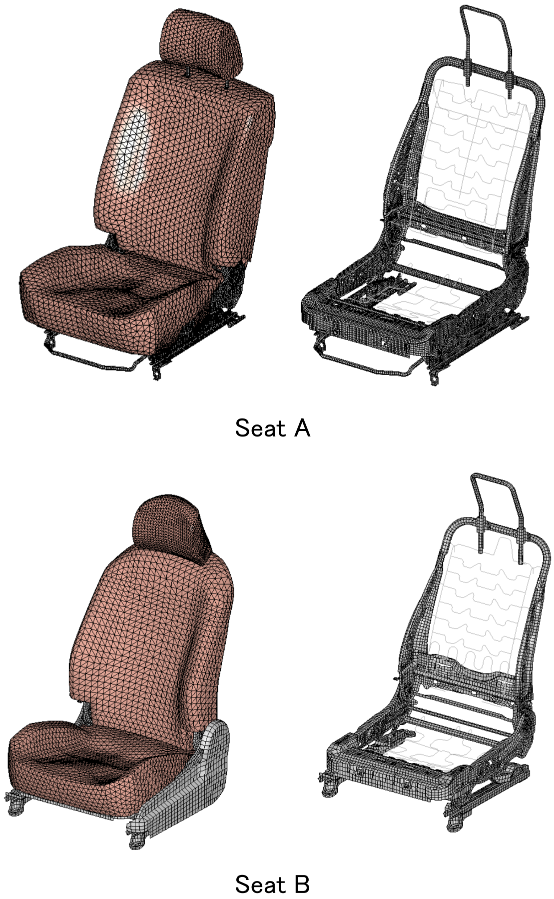


Fig.16 Seat FE model

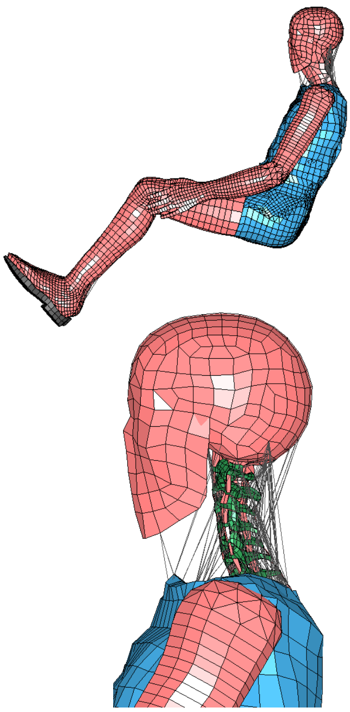


Fig.15 THUMS-AM50 OCCUPANT Ver.1.63-050304

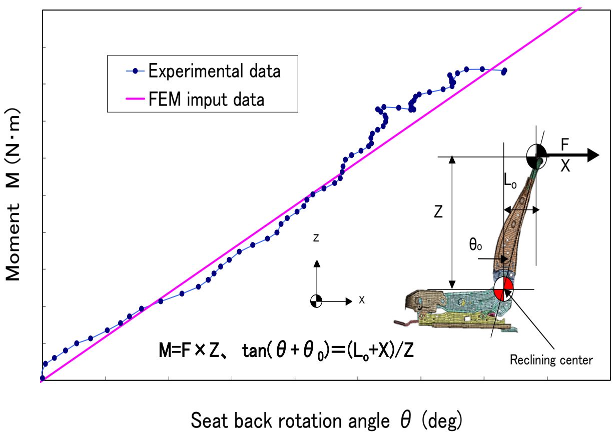
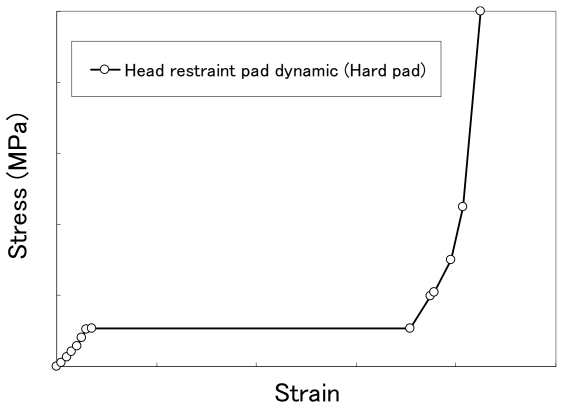
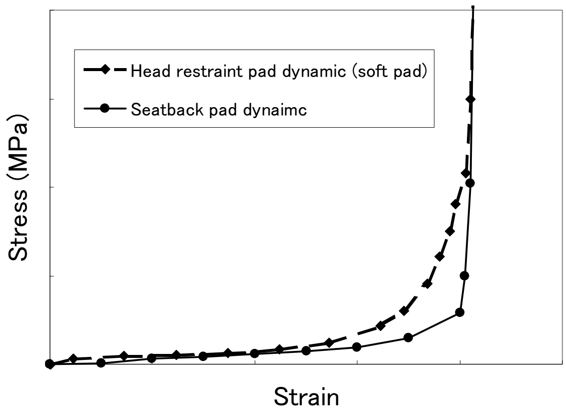


Fig.17 Seat back strength characteristic.

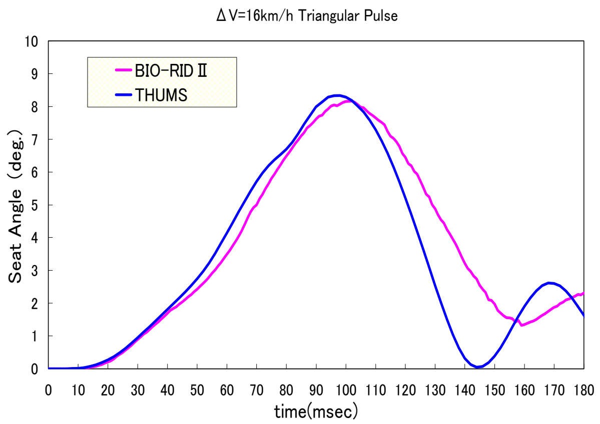


(a) Head restraint pad strength characteristic

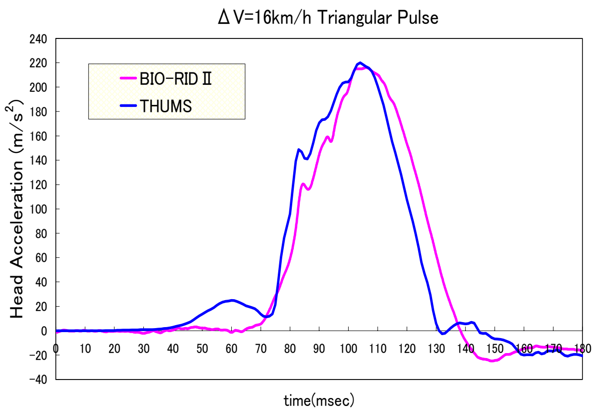


(b) Head restraint pad and seatback pad strength characteristic

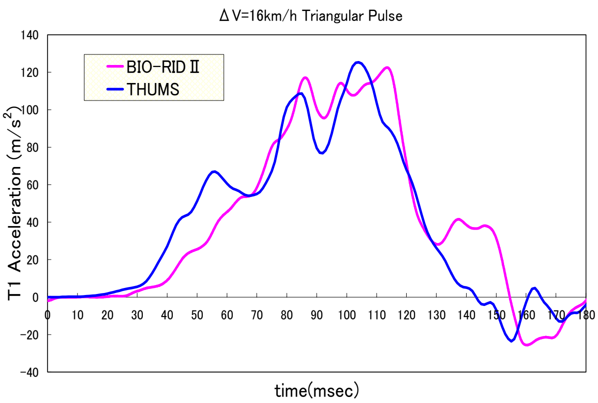
Fig.18 Head restraint and Seat back pad strength characteristic.



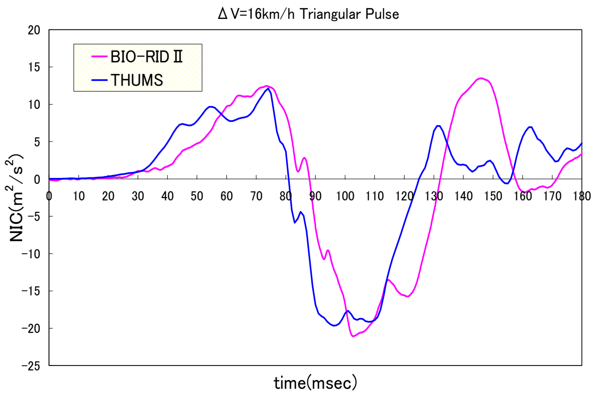
(a) Seat Angle



(b) Head Acceleration



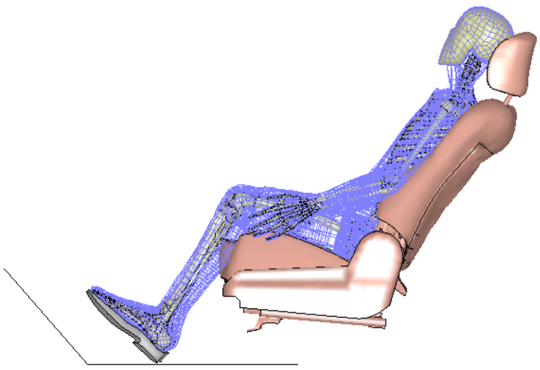
(c) T1 Acceleration



(d) NIC

Fig.19 Correspondence verification of experiment and FE analysis.

Before examination, we first made a comparison of the response of each portion between the computer models and the BIO-RID II dummy under the same conditions to verify the prediction accuracy of the models. Figure 19 shows a comparison when Seat B (the new seat) was used and pulse 1 (ΔV = 16 km/h triangular pulse) was input. With regard to the seat angle representing seat deformation, the calculation results obtained from the THUMS v1.63 and the experimental results using the BIO-RID II dummy matched well, as shown in Figure 19 (a). Additionally, the acceleration pulses and peak levels of head G and thorax G (T1G) depicting the motions of the occupant's head and neck almost matched, as shown in Figures 19 (b) and (c). A high level of consistency was also obtained for the calculated NIC, as shown in Figure 19 (d). Although we could not make any comparison with data on human subjects in this examination, the THUMS v1.63 and the seat models are considered to be highly accurate in predicting the motions of the head and the neck in rear-end collisions because their overall motions matched well with the BIO-RID II dummy, which is regarded as having high biofidelity.



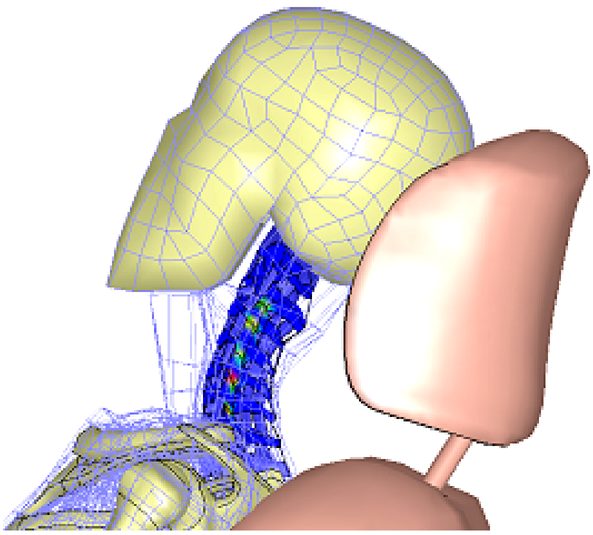


Fig.20 Example of strain distribution by FE analysis（106ms）.
(Seat B, pulse 2 : ΔV=25km/h Triangular pulse)

4.2 Results of FE analyses

Figure 20 shows the motions of the head and the neck and the strain distribution of the neck soft tissues at 106 ms after a collision. The seat B model was used for calculation, and pulse 2 (ΔV = 25 km/h triangular pulse) was input. Concerning the strain of the neck joint capsule (the soft tissue of the cervical vertebral joints) that is said to correlate with neck whiplash in rear-end collisions, it reached high at C5-C6 and C6-C7 vertebral (red area) and showed a similar tendency to the results of the volunteer tests performed by Sekizuka [15] and those of the human FE analyses conducted by Hasegawa [13].

Figure 21 shows the results of comparing neck soft tissue strain for the three input pulses between the conventional seat (Seat A) and the new seat (Seat B). The ratio of calculated strain output to the generally proposed criterion [21] was assigned to the vertical axis in the figure. The strain level was lower for the new seat than for the conventional seat for every input impulse.

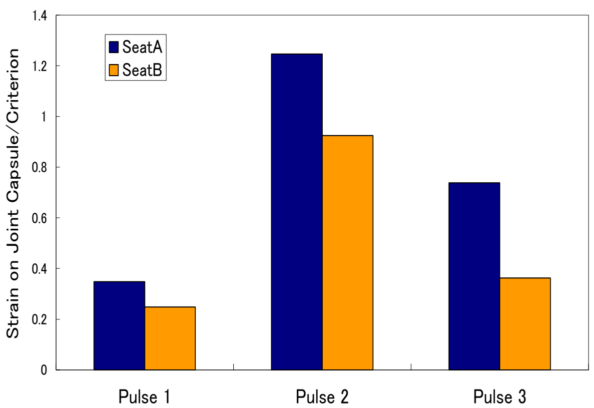


Fig.21 Strain on Joint Capsule / Criterion examination

5. CONCLUSIONS

1. We have developed a seat using the design concept of simultaneously restraining the head and thorax while preventing head rotation. The design employs higher strength seat components and an updated component layout.

2. The newly developed seat is able to simultaneously restrain the head and thorax, while minimizng the extension of the head, as intended. Measurements of simulated $T1G_{HRC}$, $T1V_{HRC}$, $T1S_{HRC}$, and NIC were used to confirm simultaneous restraint of the head and thorax; the NDC was used to measure head rotation; the LNL was used to measure load on the lower neck, and lower My_{max} was also used to ensure that the new seat could reduce whiplash injury.

3. There is a great difference in injury value even between the triangular pulse and the trapezoidal pulse at the same ΔV. In general, when a car suffers a rear-end collision G, the collision acceleration pulse seems to depend on the bullet vehicle and the wrap rate. The triangular pulse is considered to be effective in dealing more severe collision conditions.

4. In rear-end collision simulations using THUMS-AM50 OCCUPANT Ver.1.63-050304, the strain distribution of the neck soft tissues was larger in the lower neck, as observed in experiments. We also investigated the strain of the neck joint capsule, and confirmed that the strain level was lower for the new seat than for the conventional seat. This result proves that our new design is effective in reducing load on the neck. The seat development concept we proposed was proven to the effective by experiments and FE analyses.

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REFERENCES

[1] Japanese National Police Agency. 1999. ”Traffic Green Paper ’99.(Japanese)”.

[2] The General Insurance Association of Japan data. 1999.

[3] Ministry of Land, Infrastructure and Transport. 2002. “ The 3rd car safety symposium”.

[4] M. Krafft and A. Kullgren. 1998.“A. Crash Pulse Recorders in Rear Impacts Real Life Data, 16th ESV conference 1998, paper no.98-S6-O-10.

[5] O. Bostron, M. Svensson, B. Aldman, H. Hansson, Y. Haland, P. Lovsund, T. Seeman, A. Suneson, A. Saljo and T. Ortengen. 1996. “A New Neck Injury Criterion Candidate Based on Injury Findings in the Cervical Spinal Ganglia after Experimental Neck Extension Trauma.” Proc. IRCOBI Conf., pp.123-136.

[6] K.-U.Schmitt, M. H. Muser and P. Niederer. 2001. ”A New Neck Injury Criterion Candidate for Rear-end Collisions Taking into Account Shear Forces and Bending Moment.” 17th ESV conference 2001, paper no.124.

[7] F. Heitplatz, R. Sferco, P. Fat and J. Reim. 2003. “An Evaluation of Existing and Proposed Injury Criteria with Various Dummies to Determine Their Ability to Predict the Levels of Soft Tissue Neck Injury Seen in Real World Accidents.” 18th ESV conference 2003, paper no.504.

[8] D. C. Viano, S. Olsen. G. S. Lock and M. Humer. 2002. “Neck Biomechanical Responses with Active Head Restraint: Rear Barrier Tests with BioRid and Sled Tests with Hybrid III.” SAE 2002-01-0030.

[9] M. M. Panjabi, J.-L. Wang and N. Delson. 1999. “Neck Injury Criterion Based on Intervertebral Motions and its Evaluation using an Instrumented Neck Dummy.” Proc. IRCOBI Conf., pp179-190.

[10] B. Deng, P. C. Begeman, K. H. Yang, S. Tashman and A. I. King. 2000. “Kinematics of Human Cadaver Cervical Spine During Low Speed Rear-End Impacts.” 44th Stapp Car Crash Conference SAE 2000-01-SC13.

[11] K. Ono and K. Kaneoka. 1997. “Motion Analysis of Human Cervical Vertebrae During Low Speed Rear Impacts by the Simulated Sled ” Proc. IRCOBI Conf, pp223-237.

[12] S. Ejima, K. Ono and K. Yamazaki. 2004. “Analysis of Finite Element Model For Neck Injury Mechanisms in Low-Speed Rear-End Collisions.” Proc. IRCOBI Conf, pp323-324

[13] J. Hasegawa. 2004. ”A Study of Neck Soft Tissue Injury Mechanisms During Whiplash Using Human FE model.” Proc. IRCOBI Conf, pp321-322.

[14] K. E. Lee, M. E. Davis, R. M. Mejilla, and B. A. Winkelstein. 2004. “In Vivo Cervical Facet Capsule Distraction: Mechanical Implications for Whiplash and Neck Pain.” 48th Stapp Car Crash Conference SAE 2004-22-0016.

[15] M. Sekizuka. 1998. “Seat Designs for Whiplash Injury Lessening.” 16th ESV conference 1998, paper no.98-S7-O-06.

[16] B. Lundell, L. Jakobsson, B. Alfredsson, M. Lindstrom, and L. Simonsson. 1998. “The WHIPS Seat – A Car Seat for Improved Protection Against Neck Injuries in Rear End Impacts.” 16th ESV conference 1998, paper no.98-S7-O-08.

[17] K. Wiklund, and H. Larsson. 1998. “Saab Active Head Restraint (SAHR) Seat Design to Reduce the Risk of Neck Injuries in Rear Impacts.” SAE 980297.

[18] International Insurance Whiplash Prevention Group. 2004. “IIWPG Protocol for the Dynamic Testing of Motor Vehicle Seats for Neck Injury Prevention.”

[19] ADAC Home page, http://www.adac.de/

[20] M. Krafft, A. Kullgen, A. Lie and C. Tingvall. 2004. “Assesment of Whiplash Protection in Rear Impact–Crash Test and Real-life Crashes.”.,Folksam Home page ,http://www.folksam.se/

[21] N. Yoganandan, F. A. Pintar and S. Kumaresan. 1998. “Biomechanical assessment of Human Cervical Spine Ligaments.” SAE983159